Coils, Receivers, and Parallel Imaging: A Technical Perspective

Charles A. McKenzie, PhD, Daniel K. Sodickson, MD, PhD

Department of Radiology Beth Israel Deaconess Medical Center; Harvard Medical School, Boston, MA, USA

cmckenzi@bidmc.harvard.edu

Introduction

The radiofrequency (RF) receive chain, which takes an MR signal and turns it into an image, has recently seen a dramatic increase in the number of receiver channels available on clinical magnets. This talk will begin with a discussion of how coil array design and parallel imaging provided motivation for this change. It will proceed with discussion of how these newly expanded capabilities are impacting body MRI now, and what new capabilities may be opened up in the future.

Receiver Coils and Coil Arrays

In order for an MR image to be generated, the MR signal must be detected with a receive coil. This coil is a principal determinant of signal to noise ratio (SNR), and its size, shape, and positioning can exert a powerful effect on the overall appearance of MR images. Two types of receiver coils are typically used in clinical imaging: volume coils, and surface coils.

Volume coils are generally cylindrical in design and are designed to receive signal with uniform sensitivity from the entire volume of the coil. Common examples of volume coils include many head and knee coils as well as the body coil built into most clinical magnet bores. Unfortunately, the SNR of volume coils tends to decrease as the size of the coil increases. For this reason, the built in body coil generally does not have enough SNR for body imaging applications.

Surface coils, as may be inferred from their name, are designed to rest on the surface of the body part being imaged. They are most commonly designed as single loops with diameters between 3 and 20 cm. Surface coils typically provide higher SNR than volume coils, but receive signal from a limited volume directly beneath the coil. The SNR also falls off with depth, with the general rule of thumb that for any target imaging depth, a coil with diameter approximately equal to that depth will provide optimum SNR. Most importantly, the maximum SNR of a surface coil increases as the coil diameter shrinks. While surface coils can provide very high SNR, their anatomic coverage is so limited that they are not commonly used for clinical abdominal imaging.

Fortunately, the SNR advantages of small surface coils can be extended to larger fields of view by stringing RF coils together in arrays [1]. Each coil in the array provides locally high SNR, and through appropriate combination of the images from these coils, the high SNR of small coils can be extended over clinically useful FOVs. The need for high SNR in body MRI in particular has led to the extensive use of surface coil arrays specifically designed for imaging the abdomen. Imaging with arrays is routine and a wide variety of special-purpose coil arrays are currently on offer from commercial providers.

Receiver Channels

It is necessary to have access to the full complex signal from each coil in the array in order to combine the images in a manner that optimizes SNR. Simple combinations of the signal in hardware are suboptimal as they cause destructive interference of the complex signals. Independent receiver channels are required in order to reap the full benefits of imaging with coil arrays. Receiver channels are made up of various hardware components that amplify, transport, and digitize the raw MR signal from each coil. By appropriately combining the complex signals from each channel, images with optimal SNR over the entire FOV can be generated [1]. By the middle 1990's, most clinical MR scanners were equipped with between four and six receiver channels – a number which had been deemed sufficient to capture many of the SNR benefits of array imaging without incurring undue cost.

Parallel imaging

In addition to allowing large FOV imaging with high SNR, the spatial separation of the coils in an array encodes a small amount of spatial information. With proper calibration of that spatial information and subsequent combination of data acquired from the array elements, some fraction burden of spatial encoding from the gradients can be moved to the coils themselves. Multiple lines of data may thus be acquired in parallel rather than sequentially, thereby circumventing limits on imaging speed due to limited gradient switching rates.

The field of parallel imaging came to widespread attention within the MRI community with the introduction of the SMASH technique in 1997 [2], followed shortly thereafter by the introduction SENSE [3]. Since their introduction, parallel imaging has become very successful and various flavors of parallel MRI reconstructions are now available for use in routine imaging on most manufacturers' scanners.

Practical requirements of parallel imaging include the use of suitable coil arrays [4], and calibration of coil sensitivities, since the distinct "view" of each array element must be known in order to form the proper data combinations for image reconstruction. Sensitivity calibration may be accomplished using a rapid low-resolution scan covering prospective volumes of interest prior to accelerated imaging. This works well when the anatomy to be imaged and the array used for imaging remains stationary. However, in body imaging both the anatomy of interest may move from scan to scan and, since the coil array generally rests directly on the abdomen, the coils themselves may move between scans. Thus for body imaging it is desirable to calibrate the coil sensitivities with the accelerated data itself [5]. Such self-calibration is possible using appropriate data trajectories, and self-calibrating techniques such as GRAPPA [6] are commercially available today.

Because the acceleration factors that can be achieved with parallel imaging are at least partially limited by the number of coils in the array used to receive data and the design of the coil array itself, parallel imaging has had a significant impact on receive chain hardware. Coil arrays are now generally designed with parallel imaging in mind, and a number of new designs and design principles have been explored [4, 7, 8]. Additionally, motivated at least in part by parallel imaging, there has been a substantial increase in the number of available receiver channels with

several manufacturers now offering at least 32 receivers with their new scanners. In combination with many-element arrays, this will allow access to higher accelerations, as well as enabling more robust imaging at lower accelerations.

Of course, this increase in the number of receiver/coils results in as large increase in the amount of data being acquired. This is exacerbated by the increased computational demands of parallel imaging reconstructions, especially as these reconstruction algorithms are designed to handle more sophisticated and efficient data acquisition schemes [9-13].

Impact on body MRI

How will the introduction of many receiver channels and the many element arrays that they allow affect body MRI?

Obvious applications for parallel imaging, and the ones that have been primarily used up to now, include reducing acquisition time, or increasing image resolution while maintaining scan time. However, both of these strategies are necessarily limited by the cost in SNR imposed by increasing accelerations [14, 15]. These limits can be mitigated to some extent by imaging at high ($\geq$ 3.0T) field strength.

Perhaps more interestingly, parallel imaging is now opening up a variety of other strategies where acceleration of the acquisition is of secondary importance. These include:

- Improving SNR efficiency of steady state sequences [16]
- Minimizing geometric distortion of echo train acquisitions [17]
- Detecting and correcting for physiologic motion. [18-20]
- Making practical otherwise prohibitively time consuming examinations/techniques [21-23]

Implementing new and innovative techniques like these have the potential to go beyond continued incremental improvements across the spectrum of clinical imaging applications, and fundamentally change how body MRI is practiced.

References

1. Roemer, P.B., et al., *The NMR phased array.* Magn Reson Med, 1990. 16(2): p. 192-225.

2. Sodickson, D.K. and W.J. Manning, *Simultaneous acquisition of spatial harmonics (SMASH): fast imaging with radiofrequency coil arrays.* Magn Reson Med, 1997. 38(4): p. 591-603.

3. Pruessmann, K.P., et al., *SENSE: sensitivity encoding for fast MRI.* Magn Reson Med, 1999. 42(5): p. 952-62.

4. Weiger, M., et al., *Specific coil design for SENSE: a six-element cardiac array.* Magn Reson Med, 2001. 45(3): p. 495-504.

5. McKenzie, C.A., et al., *Self calibrating parallel imaging with automatic coil sensitivity extraction.* Magn Reson Med, 2002. 47: p. 529-538.

6. Griswold, M.A., et al., *Generalized autocalibrating partially parallel acquisitions (GRAPPA).* Magn Reson Med, 2002. 47(6): p. 1202-10.

7. Ohliger, M.A., et al., *Effects of inductive coupling on parallel MR image reconstructions.* Magn Reson Med, 2004. 52(3): p. 628-39.

8. Ohliger, M.A., et al., *Concentric Coil Arrays for Parallel MRI.* Magnetic Resonance in Medicine, 2005. 54: p. 1248-1260.

9. Yeh, *Inherently Self-Calibrating Non-Cartesian Parallel Imaging.* Magnetic Resonance in Medicine, 2005. 54: p. 1-8.

10. Yeh, E.N., et al., *Parallel magnetic resonance imaging with adaptive radius in k-space (PARS): constrained image reconstruction using k-space locality in radiofrequency coil encoded data.* Magn Reson Med, 2005. 53(6): p. 1383-92.

11. Pruessmann, *Advances in Sensitivity Encoding With Arbitrary k-Space Trajectories.* Magnetic Resonance in Medicine, 2001. 46: p. 638-651.

12. Sodickson, D.K. and C.A. McKenzie, *A generalized approach to parallel magnetic resonance imaging.* Med Phys, 2001. 28(8): p. 1629-43.

13. Weiger, M., K.P. Pruessmann, and P. Boesiger, *2D SENSE for faster 3D MRI.* Magma, 2002. 14(1): p. 10-9.

14. Ohliger, M.A., A.K. Grant, and D.K. Sodickson, *Ultimate intrinsic signal-to-noise ratio for parallel MRI: electromagnetic field considerations.* Magn Reson Med, 2003. 50(5): p. 1018-30.

15. Wiesinger, *Electrodynamics and Ultimate SNR in Parallel MR Imaging.* Magnetic Resonance in Medicine, 2004. 52: p. 376-390.

16. Weiger, M., et al., *Sensitivity encoding as a means of enhancing the SNR efficiency in steady-state MRI.* Magn Reson Med, 2005. 53(1): p. 177-85.

17. Griswold, M.A., et al., *Resolution enhancement in single-shot imaging using simultaneous acquisition of spatial harmonics (SMASH).* Magn Reson Med, 1999. 41(6): p. 1236-45.

18.   Atkinson, D., et al., *Coil-based artifact reduction.* Magn Reson Med, 2004. 52(4): p. 825-30.

19.   Larkman, D.J., D. Atkinson, and J.V. Hajnal, *Artifact reduction using parallel imaging methods.* Top Magn Reson Imaging, 2004. 15(4): p. 267-75.

20.   Madore, B., *UNFOLD-SENSE: a parallel MRI method with self-calibration and artifact suppression.* Magn Reson Med, 2004. 52(2): p. 310-20.

21.   Schmidt, G.P., et al., *[Comparison of high resolution whole-body MRI using parallel imaging and PET-CT. First experiences with a 32-channel MRI system].* Radiologe, 2004. 44(9): p. 889-98.

22.   Sodickson, D.K., et al., *Rapid volumetric MRI using parallel imaging with order-of-magnitude accelerations and a 32-element RF coil array: feasibility and implications.* Acad Radiol, 2005. 12(5): p. 626-35.

23.   Zhu, Y., et al., *Highly parallel volumetric imaging with a 32-element RF coil array.* Magn Reson Med, 2004. 52(4): p. 869-77.